

Instrumented Footwear Inserts: A New Tool For Measuring Forces and Biomechanical State Changes During Dynamic Movements

Joe Lacirignola¹, Christine Weston¹, Kate Byrd¹, Erik Metzger¹, Ninoshka Singh¹, Shakti Davis¹, David Maurer¹, Whitney Young¹, Paula Collins¹, Jim Balicus², Mark Richter², Jeff Palmer¹

(1) MIT Lincoln Laboratory, 244 Wood Street, Lexington, MA

(2) Marine Expeditionary Rifle Squad, Quantico, VA

Abstract—Lower-limb musculoskeletal injuries are a pervasive problem in the population and military, especially during basic training where load bearing bones and joints are repeatedly subjected to aggressive movements and high forces. The ability to measure these elements during training would be a critical enabler for prevention of injury and development of more quantitative training procedures that focus on ambulatory mobility and agility. It has been inherently difficult to acquire this data outside of the laboratory in a robust and repeatable way. Herein, we report the construction and testing of a measurement system packaged within a shoe insert that is capable of measuring forces, accelerations, rotations and elevation changes. The ability to take these measurements in a mobile system facilitates new environments to monitor complex biomechanical actions without compromising natural gait rhythms. This can result in new methods for monitoring potential changes to gait and also help with rehabilitation strategies.

Keywords—sensors; biomechanics; load; gait; ground-reaction forces; load cells

I. INTRODUCTION

Lower-limb musculoskeletal injuries are ubiquitous in the general population and especially in the military, with hundreds of thousands of injuries reported every year [1]. These injuries include acute traumas (e.g., meniscal tears) as well as overuse injuries such as stress fractures and Achilles tendonitis [2, 3]. The complex force vectors and motions that contribute to these injuries are part of a biomechanical network of lower limb movements. Lower limb biomechanics represent an examination of the interactions between joints, muscles and the nervous system that create patterns of movement and responses to external stimuli and stressors such as carrying a heavy load. Understanding the various components of the lower limb movements and the simultaneous interactions that guide their evolution requires measurement of the individual components (forces and displacements) as well as algorithms to interpret the relationship between the components.

Currently, no robust mechanism exists for quantitatively measuring lower limb biomechanics (particularly forces) on a

system level outside of the laboratory. In addition, little is known about how lower limb movement and the associated components are perturbed during unloaded or loaded activity, both prior to and after a musculoskeletal injury. Technologies to measure lower limb movement components such as ground reaction forces (GRFs) or joint accelerations have traditionally utilized lab-based, force-sensing treadmills and fixed multi-camera motion capture systems to directly quantify the kinematic and kinetic responses for static and dynamic conditions [4]. While accurate, these measurements are difficult to translate outside of the laboratory. The lack of field data limits the ability to calculate the real-time impact of environmental factors, such as different terrain types and changes in elevation, on an individual's performance or susceptibility to injury. Advances in electronics miniaturization have enabled translation of several conventional laboratory measurements to be taken within and/or near footwear [5], which presents new opportunities for ambulatory sensing [6]. Several of these technologies include different types of modified insoles such as pressure-sensitive fabrics [7], force-sensitive resistors [8], capacitive arrays [9, 27], and air pressure bladders [10]. While these approaches have limited dynamic range, sampling frequency and accuracy, they represent excellent packaging schemes and can seamlessly integrate with many types of footwear. "In-shoe" based technologies have also been developed, often using small load cells and force plates integrated around a sandal [11, 12] or underneath a warfighter boot [13]. While the in-shoe approaches tend to demonstrate higher accuracy and more robustness, they are expensive and have a larger size and weight which can alter the normal gait of the user.

Herein, we discuss components of an instrumented shoe insert technology that is integrated into a commercial insole. The load sensing elements are inexpensive, flat and flexible load cells (Figure 1). These novel load sensing elements are positioned in three locations (heel, arch, and toe) within the insert and are capable of ambulatory sensing of vertical ground-reaction forces at sampling rates of up to 600 Hz each

with a large dynamic range (up to 360 kg or 800 lbs). The instrumented insert system also has two nine-axis (three axes each for accelerometer, gyrometer, and magnetometer) motion tracking devices (MTD) at both the foot and above the ankle (Figure 2). They are constructed with readout electronics, data storage and a rechargeable battery. The instrumented inserts enable the analysis of locomotive actions without altering mobility. We utilize the system to characterize both stationary measures of weight as well as dynamic responses (e.g., during walking and running), and compare the results directly to a force-sensing treadmill. We have also demonstrated an analysis path that will allow detection of biomechanical gait state changes within prolonged exercise or training events.

II. MATERIAL AND METHODS

A. Instrumented Insert Design

We developed a novel load sensing element that consists of a circular wave spring (McMaster Carr, part number 9714K8) with a strain gage (Omega, SGT-3S/350-TY11) bonded to each of five waves (Figure 1A). A custom flexible circuit board (Figure 1B) connects the strain gages in series and provides analog to digital conversion. Three load sensing elements are packaged within a modified commercial insert (Inov-8, 6 mm footbed): one at the heel, arch, and toe areas (Figure 2). A second circuit board is located at the midsection of the insert. It contains a 9-axis motion tracking device (InvenSense, MPU-9250) and also aggregates the signals from each of the three load sensing elements and sends the signals up to ankle electronics (Figure 2). The ankle electronic unit contains a second 9-axis motion tracking device, storage, processing and a barometric pressure sensor (STMicroelectronics, LPS-25H).

The load sensors were placed in the inserts to ensure as much coverage as possible underneath the foot. The overall system goal for this prototype is to be within ± 4.5 kg (10 lbs.) of true static and dynamic weight loads.

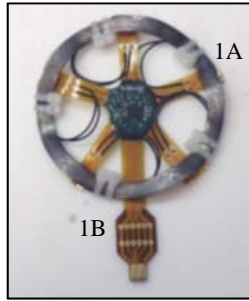


Figure 1: Custom load sensing element

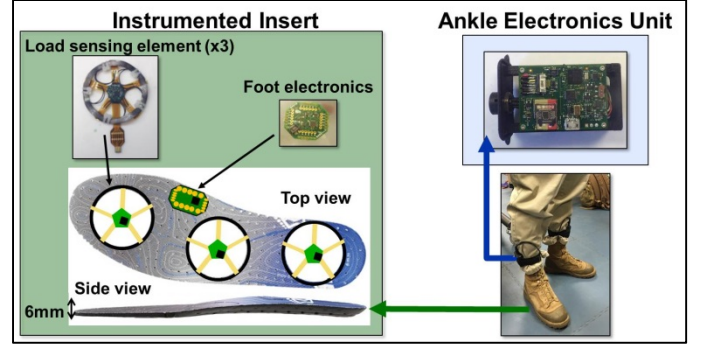


Figure 2: Construction of instrumented inserts

B. Sensor Calibration Inputs

Before starting calibration we determined the sources of variability in the system and developed a way to measure and account for the variability. The three types of variability in the system are: slight differences in load sensing element circuit components, minor nonuniformities due to manual fabrication, and end-user anthropomorphic variability (Figure 3). The variability in the load sensing circuit is due to slight variability in the performance of commercial circuit board components. To account for the circuit variability we measure a baseline value of each sensor. This baseline value is the value reported when no load is applied to the sensor. The variability in manufacturing is due to small differences in subcomponent alignments during assembly because the insert prototypes are made by hand. There are also slight variations in the insert construction. To account for manufacturing variability we measure calibration curves for each sensor and determine material properties such as creep and stress relaxation for each insert. The anthropomorphic variabilities are attributed to end-user anatomical differences and the resultant variation in weight distribution both at rest (standing, balancing) and during ambulatory movement (walking, running, climbing). We account for these different classes of variation by creating static and dynamic weight profiles.

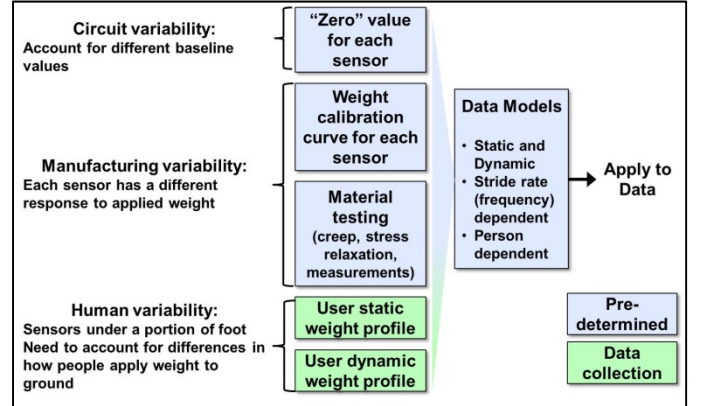


Figure 3: Sources of variability in instrumented insert

C. Sensor Calibration Curves

Each sensor in an insert is run through a sensor calibration procedure using an Instron force-displacement testing machine (Instron, 5900 series). Loads are applied in three settings: step loading, trapezoidal loading, and sinusoidal loading (Figure 4).

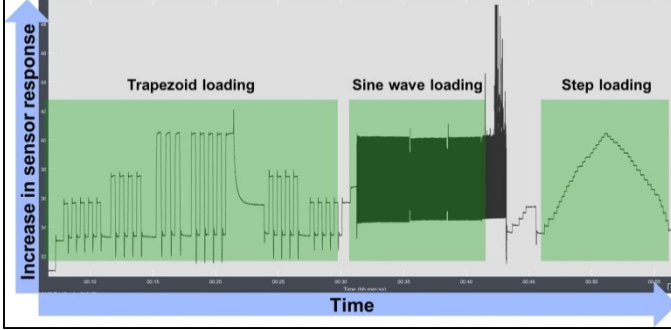


Figure 4: Instron load settings

Step loading is used to determine sensor calibration curves as well as material properties. The step testing parameters use a 4.5 to 95 kg (10-210 lbs) range, with a step size of 4.5 kg (10 lbs), a loading rate less than 1 second and a 15 second hold at each step. The sine wave loading is representative of dynamic gait loading and is used to determine dynamic loading properties. The sine loading parameters are 4.5 to 95 kg (10-210 lbs) range, with 4 different frequencies: 1, 1.5, 2 and 2.5 Hz. The cyclic sine loading lasts for a minimum of 180 complete periods. Trapezoid loading is used to determine static loading properties as well as the response to creep and stress relaxation conditions. The trapezoid parameters employ max loads of 27, 50, 95 kg (60, 110, 210 lbs) with a loading rate of either 22.5 or 45 kg (50 or 100 lbs) per second. Each max weight is held for 20 seconds followed by 20 seconds with no load. Each max load trapezoid is repeated three times.

D. Force Treadmill Collection

Insert data was collected on a Computer Assisted Rehabilitation Environment (CAREN) system split-belt force-sensing treadmill (Motek) with 10 Vicon motion capture cameras. The insert protocol was divided into a static and dynamic collection period. The static collection initiates when the user has put on boots with the insert placed inside. The user laces the boots and then lifts each foot in the air to collect a zero weight baseline on each insert, thereby also taring any compressive force from securing the foot in the boot. Next the user hops three times on the left treadmill belt to signal the start of the collection. Then starting with a single limb stance, data are collected for 30 seconds using the left leg as the support limb while on the left treadmill plate. This process was then repeated for the right single leg stance on the right treadmill plate. Next the user stands for 30 seconds while in a double limb stance (left foot on the left plate and right foot on the right plate). Then the user stands for 30 seconds with the toes of both feet on the left plate and the heels on the right plate. This is followed by hoping three times on the right

treadmill plate to complete the static collection. After a short break, the dynamic part of the collection starts with three hops on the left treadmill plate. The user then walks for one minute at two different speeds: 1.34 m/s and 1.8 m/s (3 and 4 mph) on a flat (0°) incline. The dynamic collection was signaled to be complete with a final three hops on the right treadmill plate.

E. Pre-processing: Force Treadmill and Insert Data

Pre-processing was performed using MATLAB (version R2015a). The unfiltered treadmill data (1000 Hz) was converted from newtons to pounds and then downsampled to 500 Hz using a 1D interpolation with linear sampling. Raw voltages from each load cell were acquired at a rate of 600 Hz and downsampled to 500 Hz using the same process. All treadmill and insert data were filtered using a Butterworth low-pass filter using designfilt (MATLAB, 20 Hz passband and 40 Hz stopband). This was done to reduce high-frequency noise originating from motor operation in the treadmill. Sensor data from the insert was time aligned to the force treadmill data manually using a sequence of three jumps performed at the beginning and end of each data collection trial using a user interface designed specifically for this purpose in MATLAB.

F. Calibration Proces:Static model

Each sensor within each insert requires calibration to account for sensor placement and orientation. The static calibration uses the trapezoid calibration curve generated on the Instron. Curve data was fit to a 2nd order polynomial (Figure 5). A total of six static calibration equations are generated: one for the loading and unloading properties of each load sensing element (toe, arch, and heel). Once the initial set of calibration curves are determined, they are then trained on a small subset of static data to determine the final calibration. After training, the calibration factors are used on the static data portion of the collection.

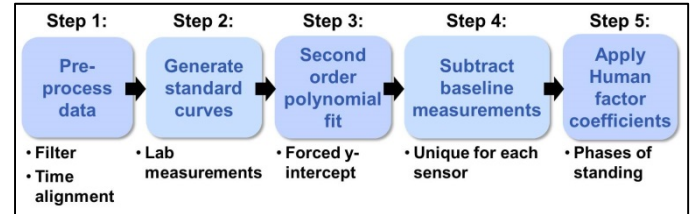


Figure 5: Steps for static model calibration

G. Calibration Proces:Dynamic Model

The dynamic calibration uses trapezoid data from each load sensing element to gain spring and damper constants. The dynamic sine wave data from the Instron testing provides initial training parameters. These parameters are then used in the viscoelastic model (modified Voight model) and applied to testing data set (Figure 6). The viscoelastic model is slightly different for each sensor to reflect differences in response based on position in the insert. The sensor data are then

summed and evaluated for fit by comparing to total force treadmill data.

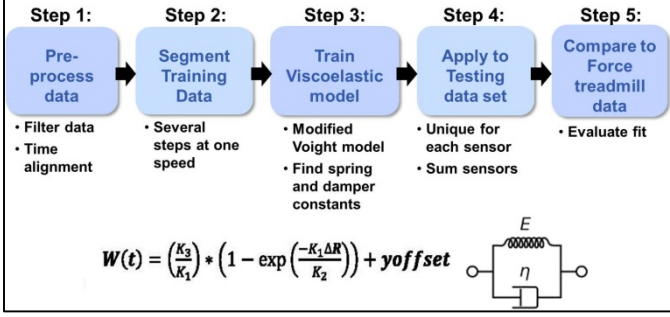


Figure 6: Steps for dynamic model calibration

H. Free Walking Data Collection

Free walking data was collected to test insert function in outdoor environments. This data consisted of walking for several minutes indoors (thin carpeted surface on concrete) and outdoors on several different surfaces (pavement, gravel and sand). The protocol consisted of adding the insert into boots, putting boots on, turning on electronics taking a baseline sensor reading (no weight) and then beginning to walk taking note of type of terrain during walk.

I. Analysis

The overarching analysis for the instrumented insert begins with inputs from each of four sensor modalities: the barometric pressure sensor, the foot MTD, the ankle MTD and the load sensing elements (Figure 7).

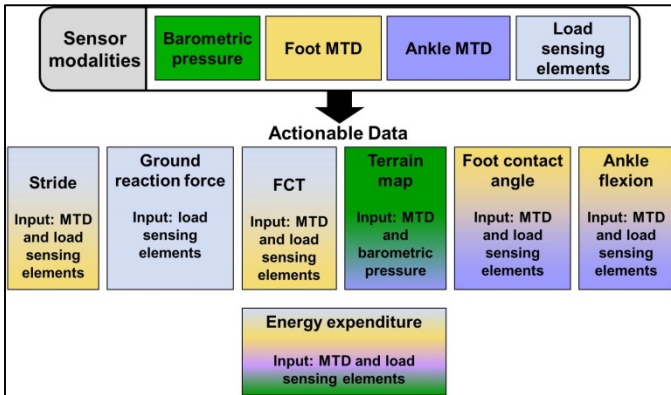


Figure 7: Overview of instrumented insert analysis structure

For the purposes of an initial analysis, only stride, GRFs and FCT were calculated using the load sensing elements (foot MTD was used as a reference). Individual strides were identified from the downsampled and filtered data. First, a 10-point, zero-phase boxcar smoothing was applied to the downsampled data and then normalized to zero mean and unit variance. This smoothed the data sufficiently to ensure peak-finding accuracy. Strides were identified by cross-correlating a “stride template,” (a sum of three Gaussians) to simulate an ideal double-peak stride representing heel strike and toe-off (Figure 8). Once strides were segmented from the peaks in the

cross-correlation function, exact timing and values for force peaks were determined using peak detection on the normalized and averaged force data, and then translated to actual force values and timings. The calculation for each step includes: duration of each step, FCT per step, percent of time spent walking (versus static data), number of steps, cadence (steps per minute), first (heel strike) and second peak (toe off), percent time in stance phase and swing phase, and information on how weight is distributed between the three load sensing elements in each insert.

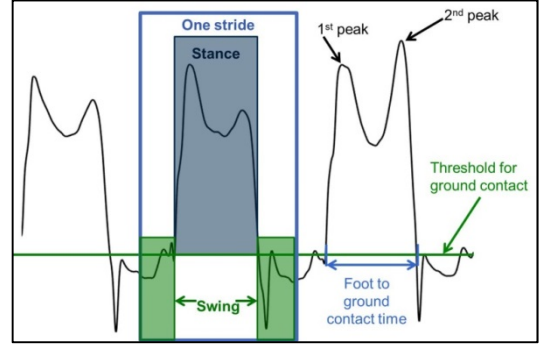


Figure 8: Stride analysis

After load and stride information is calculated, the data can then be divided into activity segments. Activity segments are determined by comparing periods of dynamic activity followed by periods of inactivity (work-rest cycles) (Figure 9).

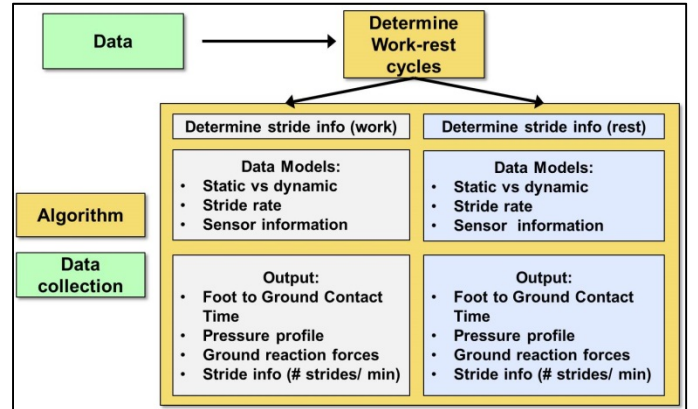


Figure 9: Determination of work-rest cycles and stride calculations

III. RESULTS

A. Preliminary Force Treadmill Comparison: Static Data

For the static data, the summed weight from the calibrated instrument insert was compared to the static force treadmill data. The static data collection is shown in Figure 10. First the subject stood on the right foot (single limb stance) with a mean force on the right insert of 88.0 with a bias of ± 0.6 kg (194.61 \pm 1.33 lbs.). Next the subject is standing on the left foot (single limb stance) with a mean force of 89.2 and a bias of ± 1.61 kg (197.37 \pm 3.57 lbs.). Finally, the subject stood on both feet (double limb stance) with a mean force on the left foot of 46.4 bias of ± 2.38 kg (102.35 \pm 5.22 lbs.) and a

mean force on the right foot of 41.7 with a bias of ± 1.4 kg (92.08 \pm 3.1 lbs.).

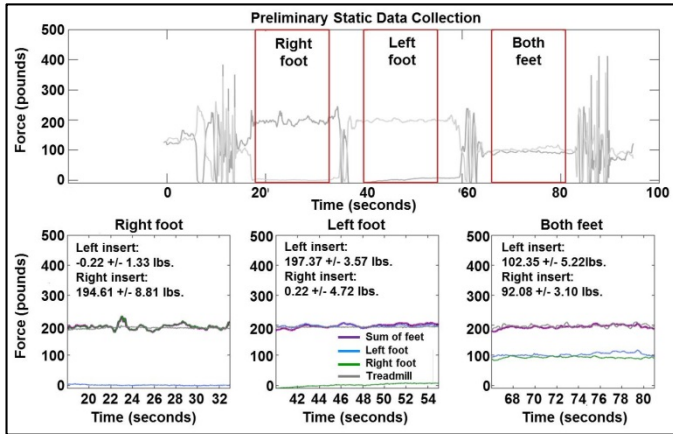


Figure 10: Preliminary Static Data

B. Preliminary Force Treadmill Comparison: Dynamic Data

For the dynamic data, the summed weight from the calibrated instrument insert was compared to the dynamic portion of the force treadmill data. An example of dynamic data are shown in Figure 11. These data are from a subject walking at 1.34 and then at 1.8 m/s (3.0 and 4.0 mph). The 1.34 m/s dynamic data shows a mean force of 97 with a bias of ± 3.62 kg (215.0 \pm 7.98 lbs.) on the 1st peak (rear heel strike) followed by a mean force of 89.8 and a bias of ± 6.5 kg (198.1 \pm 14.32 lbs.) for the 2nd peak (toe off) for both the left and right insert. The 1.8 m/s dynamic data shows a mean force of 111 with a bias of ± 3.7 kg (245.3 \pm 8.23 lbs.) on the 1st peak (rear heel strike) followed by a mean force of 93.4 and a bias of ± 6.26 kg (206.7 lbs. \pm 13.8 lbs.) for the 2nd peak (toe off) for both the left and right insert.

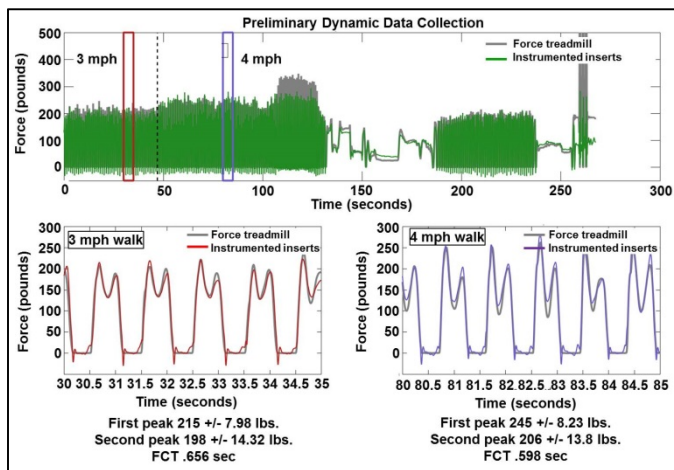


Figure 11: Preliminary Dynamic Data

C. Instrumented Insert Free Walking Data Comparison

Free walking data was also collected from the instrumented insert (Figure 12). The walk began on level ground and showed a mean first peak of 90.1 kg (196.0 lbs.) and a mean second peak of 97.2 kg (214.2 lbs.) with a mean FCT of 0.667 seconds for both the left and right insert.

Later in the walk the subject goes downhill and shows a mean first peak of 102 kg (226 lbs.) and a mean second peak and 114 kg (250 lbs.) with a lower mean FCT of 0.581 seconds for both the left and right insert.

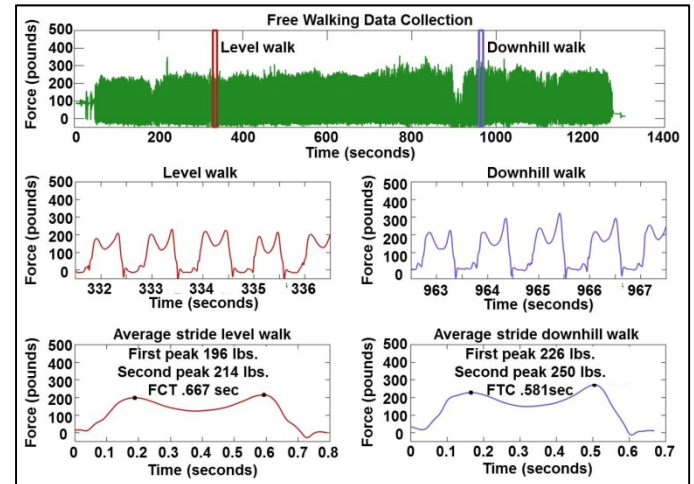


Figure 12: Example free walking data

D. Load Transition During Ground Contact

During normal walking gait, load is transitioned from the heel through the mid-foot to the toe. Since the instrumented insert has three load sensing elements positioned at the heel, arch and toe, it can track the progression of load as it travels across the foot. It can also track changes in loading shape and load duration for each sensor. Tracking this transition of load can determine if dynamic biomechanical load responses are changing over the duration of an activity or condition. In Figure 13, example load transition data are shown during the free walking data collection. Two segments are chosen for comparison, one walking on level ground and one walking downhill, to determine if the loading shape is consistent throughout the walk. The force is normalized to the level walking segments. There is a slight increase in the height of the second peak when the subject walks downhill. Downhill walking has a 0.8 seconds decrease in FCT from level walking. The difference in FCT between level and downhill walking is seen at the transition between the heel and arch sensors, providing a connection between internal biomechanical responses and changing external environmental loading conditions.

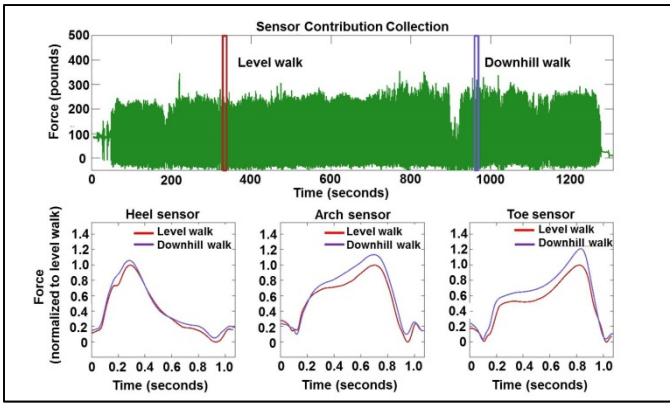


Figure 13: Load transitioning during ground contact

E. State Measurements and Comparisons

The long-term goal for comparing instrumented insert data and understanding changes in gait is to determine features that characterize load, as well as foot and ankle movement dynamics, and in conjunction with multiple multivariate regression techniques, map the state space for gait anomalies.

To begin to understand which measurable features are important for the gait anomaly state space we looked at radar plots which have the capability of showing multiple measurements in a 2D plot. Figure 14 shows the distribution of a subject walking on a treadmill at 3 mph, the distribution from the same subject walking outdoors on pavement at 3 mph and a comparison of treadmill to outdoor walking. There is an increase in FCT of over 30% while walking on treadmill as compared to walking on pavement. While walking on pavement there is a modest increase in mean percent time in stance and mean toe off with a modest decrease in mean swing percentage and mean heel strike.

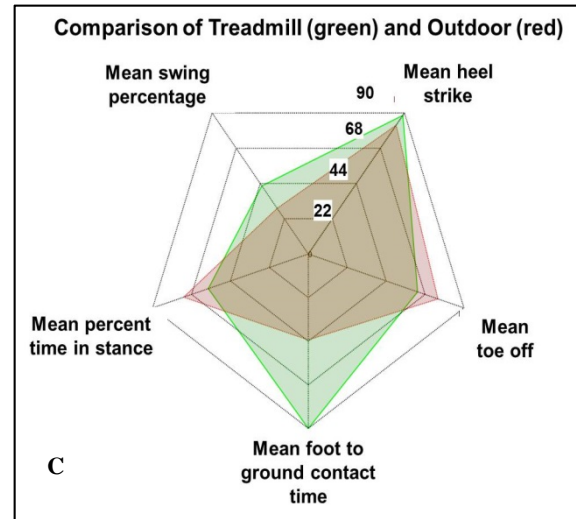
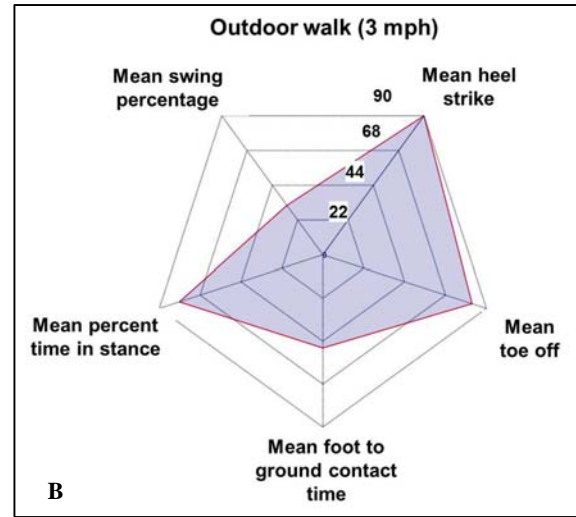
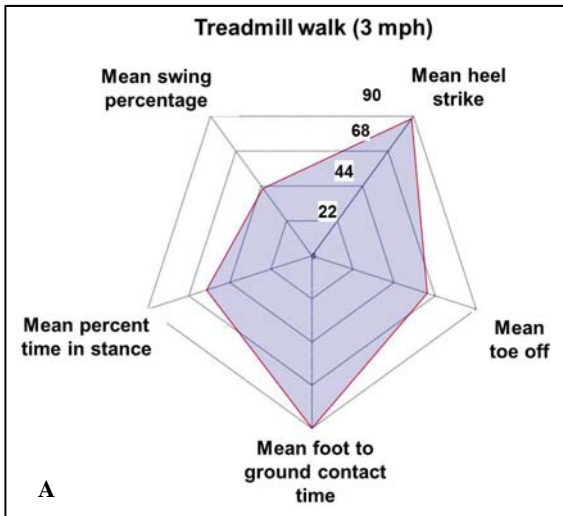


Figure 14: State measurement comparisons between walking conditions: A) Treadmill walking (3 mph), B) Outdoor walk on pavement (3 mph), C) Comparison of A and B (green is treadmill, while red is outdoor).

IV. DISCUSSION

A. Accuracy-force treadmill comparison

The preliminary force treadmill comparison shows that the inserts are at or near the goal of ± 4.5 kg (10 lbs.) for ambulatory weight measurements. We believe the dynamic toe off measurements are slightly above the goal due to sensor placement and can be corrected through human factor adjustments. The initial calibration procedure shows promise with additional work concentrating on adjusting the viscoelastic model to better represent human foot distribution of total weight. Even with accuracy at its current level, the instrument insert holds value in beginning to understand lower limb biomechanics movement, particularly extending lab capabilities to free living environments and real-time applications.

B. Free walking data

The free walking data holds a wealth of information about a person's gait. Initially we looked at a few measurements based solely on the load sensing elements: heel and toe peaks, time in stance verse time in swing and foot to ground contact time. By comparing GRF and FCT data from different environmental conditions (e.g., level vs. downhill walking) there are clear differences, even in a relatively short collection. By looking closer at how individual components respond to different activities we may be able to gain insight into how lower limb biomechanics respond to stressors and look for signs of gait abnormality.

We continue to improve the models and calibration procedures. Eventually this will become simpler as the load sensing elements and insert construction become more uniform under manufacturing conditions. We are working with several manufacturing companies to improve the insert build process to achieve better reproducibility over the current hand assembly procedure.

C. Future work

We have built an instrumented footwear insert that is capable of measuring individual components of lower limb biomechanics movement. The instrumented inserts are also a comprehensive tool for gaining insight into how lower limb biomechanics change with respect to varying environmental conditions. With more field collections, we hope to start uncovering relationships that allow understanding of how and why injuries occur, as well as mitigation strategies.

There are several user communities that are interested in the type of data measured by the instrumented inserts. These include the military (dismounted and training), rehabilitation personnel and physical therapists, prosthetics and robotics developers, athletes and users interested in sports and fitness and emergency first responders. These user communities all need to understand the impact of load on the human during complex gait actions and develop methods and procedures to mitigate injury. Ambulatory load data could help quantify performance for mobility and agility constructs and enhanced training and operational tasks [29], aid in understanding the impact of wearable equipment, help determine the metrics for enhanced running performance [26] and gain insight into the gait anomaly state space [28]. A modified version of the force sensing elements could also be used during rehabilitation to aid design, fitting and monitoring of assistive devices. Lastly, the instrumented insert could be integrated with existing heart rate sensing to inform a real-time energy expenditure calculation [30, 31] for a variety of ambulatory tasks.

ACKNOWLEDGMENTS

The authors would like to thank Todd Mower, Keri Mroszczyk, Jim Williamson, and Paul Calamia, of MIT Lincoln Laboratory, and Dave Aubin (formerly MIT Lincoln Laboratory). This work was supported by the Marine Expeditionary Rifle Squad

DISTRIBUTION STATEMENT A (Pending Approval).
Approved for public release: distribution unlimited.

This material is based upon work supported under Air Force Contract No. FA8721-05-C-0002 and/or FA8702-15-D-0001. Any opinions, findings, conclusions or recommendations expressed in this material are those of the author(s) and do not necessarily reflect the views of the U.S. Air Force.

REFERENCES

- [1] Hauret, K.G., Jones, B.H., Bullock, S.H., Canham-Chervak, M. & Canada, S. Musculoskeletal injuries: description of an under-recognized injury problem among military personnel. *Amer J. Prev. Med.* 38, 61-70 (2010).
- [2] Yassi, A. Repetitive strain injuries. *Lancet* 349, 943-947 (1997).
- [3] Milner, C.E., Ferber, R., Pollard, C.D., Hamill, J. & Davis, I.S. Biomechanical factors associated with tibial stress fracture in female runners. *Med. Sci. Sports. Exerc.* 38, 323-328 (2006).
- [4] Davis, R.B. Clinical gait analysis. *IEEE Eng. Med. Biol. Mag.* 7, 35-40 (1998).
- [5] Liedtke, C., Fokkenrood, S.A., Menger, J.T., van der Kooij, H. & Veltink, P.H. Evaluation of instrumented shoes for ambulatory assessment of ground reaction forces. *Gait & Posture* 26, 39-47 (2007).
- [6] Tao, W., Liu, T., Zheng, R. & Feng, H. Gait analysis using wearable sensors. *Sensors* 12, 2255-2283 (2012).
- [7] Razak, A.H.A., Zayegh, A., Begg, R.K. & Wahab, Y. Foot plantar pressure measurement system: a review. *Sensors*, 12, 9884-9912 (2012).
- [8] Bamberg, S.J.M., Benbasat, A.Y., Scarborough, D.M., Krebs, D.E. & Paradiso, J.A. Gait analysis using a shoe-integrated wireless sensor system. *IEEE Trans. Inform. Tech. in Biomed.* 12, 413-423 (2008).
- [9] Putti, A.B., Arnold, G.P., Cochrane, L. & Abboud, R.J. The pedar inshoe pressure system: repeatability and normal pressure values. *Gait & Posture*, 25, 401-405 (2007).
- [10] Kong, K. & Tomizuka, M. A gait monitoring system based on air pressure sensors embedded in a shoe. *IEEE Trans. Mechatr.* 14, 358-370 (2009).
- [11] Veltink, P.H., Liedtke, C.B., Droog, E. & van der Kooij, H. Ambulatory assessment of ground reaction forces. *IEEE Trans. Neur. Syst. & Rehab. Engin.* 13, 423-427 (2005).
- [12] Schepers, H.M., Koopman, H.F.J.M. & Veltink, P.H. Ambulatory assessment of ankle and foot dynamics, *IEEE Trans. Biomed. Eng.* 54, 895-902 (2007).
- [13] Lind, R.F., Love, L.J., Rowe, J.C., Pin, F.G. Multi-axis foot reaction force / torque sensor for biomedical applications. *IEEE / RSJ Int. Conf. Intell. Robots Syst.* St. Louis, MO, USA, Oct. 11-15, 2009.
- [14] Najafi, B., Khan, T., Wrobel, J. Laboratory in a box: wearable sensors and its advantages for gait analysis. *IEEE Engin. Med. Biol. Soc. Ann. Int. Conf.* Boston, MA, USA, 6507-6510 (2011).
- [15] Clarke, T.E. The pressure distribution under the foot during barefoot walking. Ph.D. Thesis, The Pennsylvania State University, University Park, PA (1980).
- [16] Cavanaugh, P.R. & LaFortune, M.A. Ground reaction forces in distance running. *J. Biomech.* 13, 397-406 (1980).
- [17] Faivre, A., Dahan, M., Parratte, B. & Monnier, G. Instrumented shoes for pathological gait assessment. *Mech. Res. Commun.*, 31, 627-632 (2004).
- [18] Lieberman, D.E. et al. Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature* 463, 531-535 (2010).
- [19] Kluitenberg, B., Bredeweg, S.W., Zijlstra, S., Zijlstra, W. & Buist, I. Comparison of vertical ground reaction forces during overground and treadmill running. A validation study. *BMC Musculo. Dis.* 13, 1471-2474-13-235 (2012).
- [20] Zachazewski, J.E., Riley, P.O. & Krebs, D.E. Biomechanical analysis of body mass transfer during stair ascent and descent of healthy subjects. *J. Rehab. Res. Develop.* 30, 412-422 (1993).

- [21] Protopapadaki, A., Drechsler, W.I., Cramp, M.C., Coutts F.J. & Scott, O.M. Hip, knee, ankle kinematics and kinetics during stair ascent and descent in healthy young individuals. *Clin. Biomech.* 22, 203-210 (1993).
- [22] Haslam, R.A. et al. The effect of military load carriage on ground reaction forces. *Gait & Posture* 26, 611-614 (2007).
- [23] Sell, T.C. et al. Minimal additional weight of combat equipment alters air assault soldiers' landing biomechanics. *Military Med.* 175, 41-47 (2010).
- [24] Sell, T.C. et al. The addition of body armor diminishes dynamic postural stability in military soldiers. *Military Med.* 178, 76-81 (2013).
- [25] Hamner, S.R., Seth, A. & Delp, S.L. Muscle contributions to propulsion and support during running. *J. Biomechanics* 43, 2709-2716 (2010).
- [26] Clark, Kenneth P., Laurence J. Ryan, and Peter G. Weyand. "Foot speed, foot-strike and footwear: linking gait mechanics and running ground reaction forces." *The Journal of experimental biology* (2014): jeb-099523.
- [27] Stöggl T, Martinier A "Validation of Moticon's OpenGo sensor insoles during gait, jumps, balance and cross-country skiing specific imitation movements." *Journal of Sports Science* 35, 196-206 (2017)
- [28] James R. Williamson, J.R., Dumas, A., Ciccarelli, G., Hess, A. R., Buller, M.J. and Telfer, B.A. "Estimating Load Carriage from a Body-worn Accelerometer" BSN 2015 conference paper.
- [29] Zaferiou, Antonia, Lauro Ojeda, Stephen Cain, Rachel Vitali, Steven Davidson, Leila Stirling, and Noel Perkins. "*Using Inertial Measurement Units Mounted on the Feet to Define Performance Metrics for an Outdoor Agility Drill.*" In *40th Annual Meeting of the American Society of Biomechanics*. Raleigh, NC, 2016.
- [30] Pandolf KB, Givoni B, Goldman RF. Predicting energy expenditure with loads while standing or walking very slowly. *J Appl Physiol* 43: 577–581, 1977.
- [31] Epstein, Y., Stroschein, L. A., Pandolf, K. B. Predicting metabolic cost of running with and without backpack loads. *European Journal of Applied Physiology and Occupational Physiology*. September 1987, Volume 56, Issue 5, pp 495–500